

# An Investigation of Three types of Tooth Implant Supported Fixed Prosthesis Designs with 3D Finite Element Analysis

Sara Koosha<sup>1</sup>, Fatemeh Sadat Mirhashemi<sup>2</sup>

<sup>1</sup>Assistant Professor, Department of Prosthodontics, Islamic Azad University of Medical Science, Branch of Dentistry, Tehran, Iran  
<sup>2</sup>Mechanical Engineer, Iran Center for Dental Research, Shahid Beheshti University M.C, Tehran, Iran

## Abstract

**Objective:** Tooth/implant supported fixed prostheses may present biomechanical design problems, as the implant is rigidly anchored within the alveolus, whereas the tooth is attached by the periodontal ligament to the bone allowing movement. Many clinicians prefer tooth/implant supported fixed prosthesis designs with rigid connectors. However, there are some doubts about the effect of attachment placement in different prosthesis designs. The purpose of this study was to examine the stresses accumulated around the implant and natural teeth under occlusal forces using three dimensional finite element analysis (3D FEA).

**Materials and Methods:** In this study, different connection designs of tooth/implant fixed prosthesis in distal extension situations were investigated by 3D FEA. Three models with various connection designs were studied; in the first model an implant rigidly connected to an abutment, in the second and third models an implant connected to abutment tooth with nonrigid connector in the distal part of the tooth and mesial part of the implant. In each model, a screw type implant (5×11mm) and a mandibular second premolar were used. The stress values of these models loaded with vertical forces (250N) were analyzed.

**Results:** There was no difference in stress distribution around the bone support of the implant. Maximum stress values were observed at the crestal bone of the implant. In all models, tooth movement was higher than implant movement.

**Conclusion:** There is no difference in using a rigid connector, non rigid connector in the distal surface of the tooth or in the mesial surface of an implant.

**Key Words:** 3D Finite Element Analysis; Fixed Prosthesis; Rigid Connection; Non- Rigid Connection

Corresponding author:  
S. Koosha, Department of  
Prosthodontics, Azad University  
of Medical Science, Tehran,  
Iran

koosha\_sa2003@yahoo.com

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## INTRODUCTION

Dental implant has been accepted as a successful clinical reality due to osseointegration. Despite the fact that dental implants have been

used widely for restoring complete and partial edentulous jaws, still there is debate on connecting the implant to the natural tooth [1].

Having used implant in fixed partial denture, fixed prostheses can be supported either by implant or tooth/implant (tooth/implant supported fixed prostheses [TIFPs]).

According to Branemark protocol in partially edentulous dentition taking into account the differential reaction of the implant and natural tooth in static and dynamic load, the connection of the natural tooth to the implant is avoided [2]. Since there is a biomechanical challenge in connecting teeth to osseointegrated implants, the use of rigid connectors (RCs) in TIFPs is not supported by Skalak [3,4]. Implants being rigidly fixed to the bone differ from natural teeth surrounded by periodontal ligaments in terms of viscoelastic properties. Consequently, under masticatory load, different patterns of stress and strain can be seen in the bone around the implant and the tooth [5-8]. Besides, the stress and strain pattern on natural teeth and the surrounding structures, which healthy periodontal ligament allows a mobility of 50 to 200  $\mu\text{m}$  to the natural teeth and the flexibility of the bone may allow 10  $\mu\text{m}$  implant movement [6]. The other criterion is that no lateral force should be designed on the prosthesis.

To compensate for dissimilar mobility of natural teeth and implant systems, several specific

methods have been suggested.

Non rigid connectors (NRCs) act as stress breakers with the ability to separate the splinted units [8,14], an implant with a stress-absorbing element (intra mobile element or stress-breaking element) or an implant with a stress-eliminating space have been recommended by some authors for TIFPs [6,15,20]. However, several reports have explored the use of non-rigid connectors and the association with abutment tooth intrusion [16,17].

Theoretically, the tooth intrusion phenomenon could be the consequence of disuse atrophy, mechanical binding and weakened rebound memory.

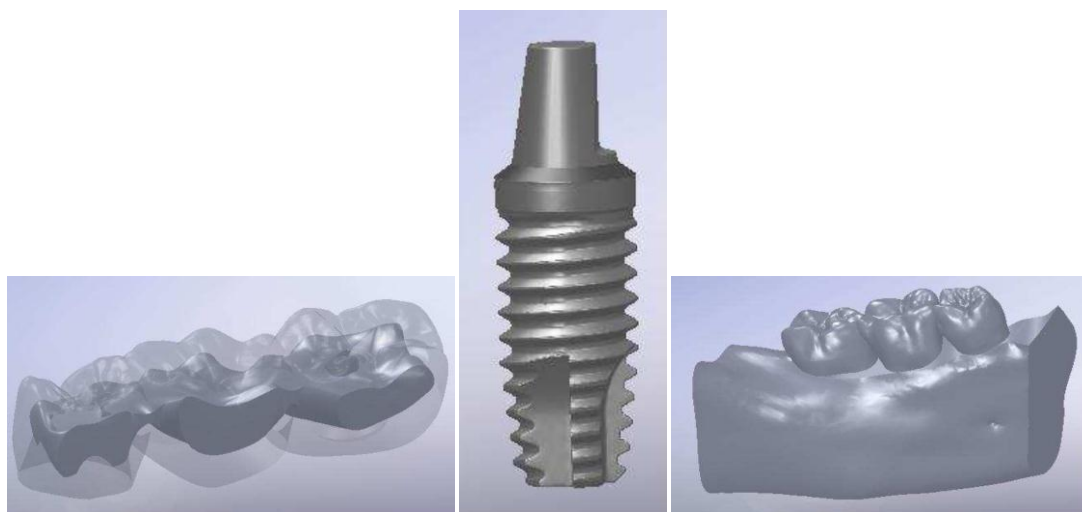
On the other hand, thanks to prosthesis and implant, rigid connectors have the inherent flexibility to modify dissimilar mobility characteristics [18].

Although the long-term radiographic evaluation of TIFPs is in favor of non-rigid connectors for less bone loss around the implant in comparison with rigid connectors [18,19], there is no agreement on proper connector's design selection for TIFPs systems.

The outcome of in vitro studies has shown unequal force distribution which is not usually compatible with the observed results in in vivo studies [20-23, 28-33].

**Table1.** Mechanical properties of the Materials

Material	Young (Elastic)'s Modulus (MPa)	Poisson's Ratio
Cortical bone	13700	0.30
Cancellous bone	1370	0.30
Titanium (implant system)	103400	0.35
Gold alloy	100000	0.30
Porcelain	69000	0.28
Dentin	18600	0.31
Pulp	3	0.45
Periodontal ligament	69	0.45



**Fig1.** The prepared 3D solid model, A) the 3-unit FP, B) the implant system, C) the assembled model consisting of tooth (second premolar), PDL, the implant system, the 3-unit FP, and alveolar bone.

## MATERIALS AND METHODS

The review of the literature reveals that consensus regarding the use of RCs, NRCs or stress-absorbing element has not been achieved. Besides, there is no detailed assessment of the role and location of NRCs between the natural tooth and implant [24-27, 34]. In this in vitro study, our null hypothesis was that various connection designs in the TIFPs may not change the load transfer between the implant and tooth abutments. Therefore, the purpose of this study was to examine the stress distribution on the supporting structures of the TIFPs under static vertical loads with the 3D FEA. In this study, three types of different TIFP designs in the distal extension partially edentulous mandible were evaluated. It was assumed that the first and second molars were extracted and an implant was inserted in the second molar position. Computer tomography (CT) images of an adult human mandible was used to make the three dimensional model of the edentulous mandible distal to the second premolar [35-37].

CT data in DICOM format was imported into the software of Rapid Form (INUS Technology, Seoul, Korea).

In this software, the CT data were directly converted to surface and finally the solid model. The height of the posterior mandibular region was determined as 23mm, the cortical bone thickness was determined as 2mm and the periodontal membrane width was accepted as 0.25mm [38].

For the solid-model construction of these prostheses, three extracted intact teeth (second premolar, first and second molars) and an implant fixture (Biomet 3i, 5×11) were used.

They were digitized using the optical digitizing system ATOS II (GOM, Braunschweig, Germany) [39,41].

This system digitizes the objects with high accuracy and 3D local. The measured data can be exported as point clouds, sections or STL-data. Here, STL-data imported to Rapid Form (INUS Technology, Seoul, Korea) was used to make the solid models of the teeth and implant system (Fig 1) [41-,43].

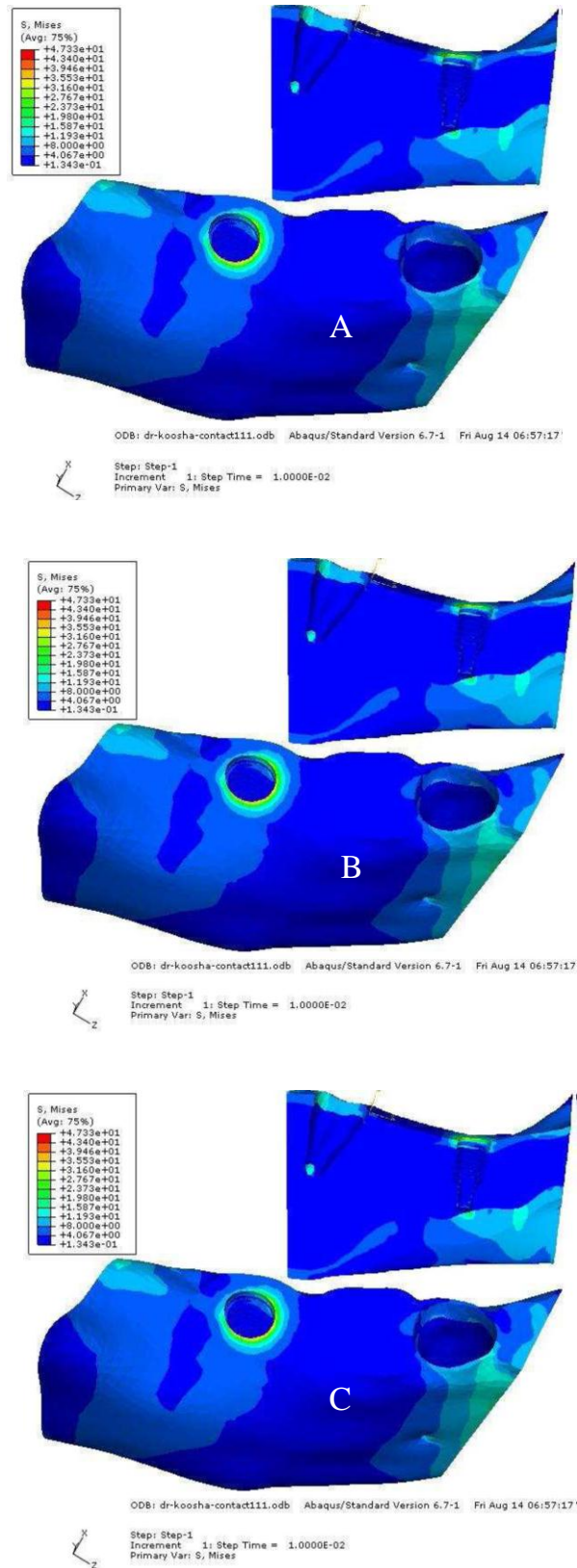


Fig2. The equivalent Von Mises stress contours, A) model 1, B) model 2, C) model 3

Then the models of the pulp and the simplified 0.25mm periodontal ligament (PDL) of the second premolar were obtained as reference from Wheeler's measurements [38]. The axes of the natural teeth and the implants in the models were compatible with the Spee Curve. In addition, the prepared models of the teeth were used to construct a 3-unit FPD in Solid Works 2008 environment [42].

In this design of the prosthesis, the thickness of porcelain was 1.5 to 2mm and gold alloy was used as a metal substructure material (Table 1) [44]. Finally, all these prepared models were assembled. Three models consisted of:

**Model 1:** The second premolar and the implant were connected rigidly.

**Model 2:** The second premolar and the implant were connected by a non-rigid attachment with the matrix connector positioned on the distal side of the second premolar.

**Model 3:** The second premolar and the implant were connected by a non-rigid attachment with the matrix connector positioned on the mesial side of the implant (Table 2).

When a rigid connector was used in the FE models, the nodes attached to the patrix/matrix components at the same location needed to merge to modify the original interfacial fixation (contact), becoming a bonded condition.

This did not allow relative micro-motion and the displacement was continuous between the different materials.

In this study, mesh generation and data processing were carried out in the 3D FEM analysis package (ABAQUS V6.7-1; Simulia Corp., Providence, USA). The mesh consisted of the 4-node linear tetrahedral solid elements with an approximate element size of 0.3mm (300 $\mu$ m) to obtain more accurate results [38]. The entire model included 911,449 elements and 1,196,657 nodes.

Materials used in this study were evaluated as homogenous, isotropic and linear and the osseointegration of the implants was accepted as 100% [26]. In the mathematical model, while the implants were directly in contact with the bone, the natural teeth had primary mobility within the borders of the periodontal membrane. Besides, the matrix and the patrix surfaces of the NRC of the TIFP were allowed to vertically move on each other. The nodes at the mesial and distal surfaces of the alveolar bone were fixed in all directions as the boundary condition. Contact between the patrix and matrix surfaces of the nonrigid connectors was assigned as tangential-frictionless (non bonded), to simulate the sliding function of a non-rigid connector; whereas the other parts of the model were assumed to be completely tied to each other. A linear static analysis was performed on the prepared 3D solid models with a vertical occlusal load of 250 N on the occlusal surface of each tooth at a right angle (0° to the long axis of supports) on the central fossa.

**Table 2.** Fixed Prosthesis Designs Used in this Study

Fixed prosthesis design	
<b>Model 1</b>	The second and the implant are connected rigidly
<b>Model 2</b>	The second premolar and the implant are connected by a nonrigid attachment with the matrix connector positioned on the distal side of the premolar
<b>Model 3</b>	The second premolar and the implant are connected by a non-rigid attachment with the matrix connector positioned on the mesial side of the implant

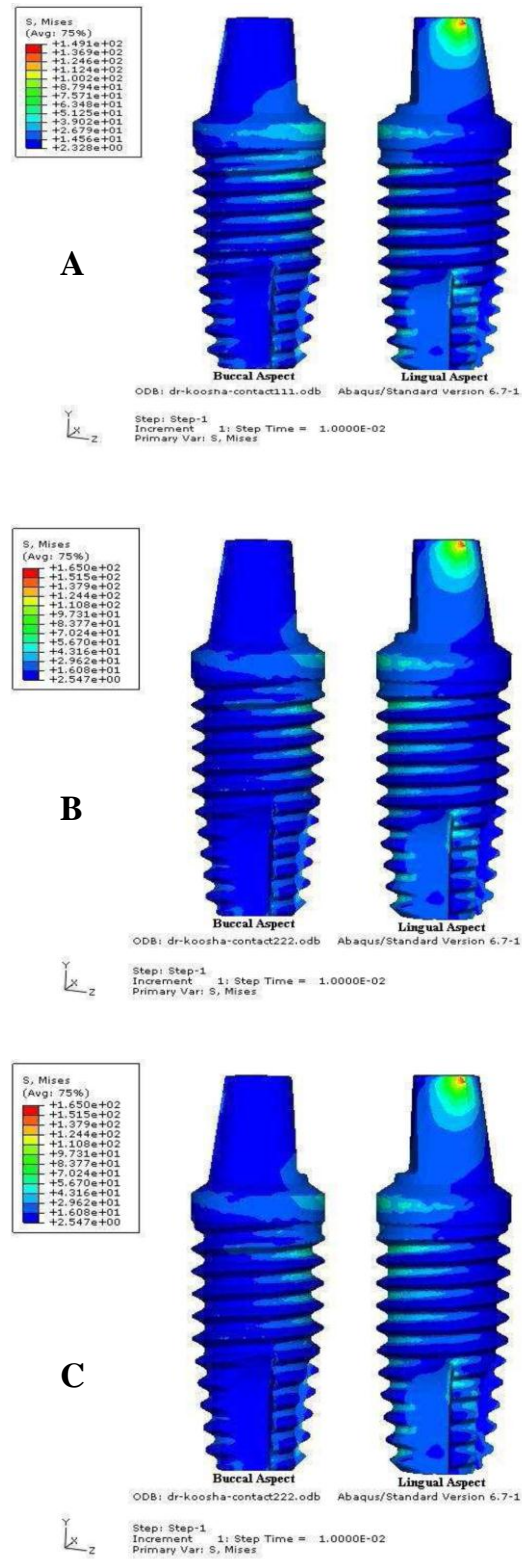


Fig 3. The Von Mises stress contours of the implant, A) model 1, B) model 2, C) model 3

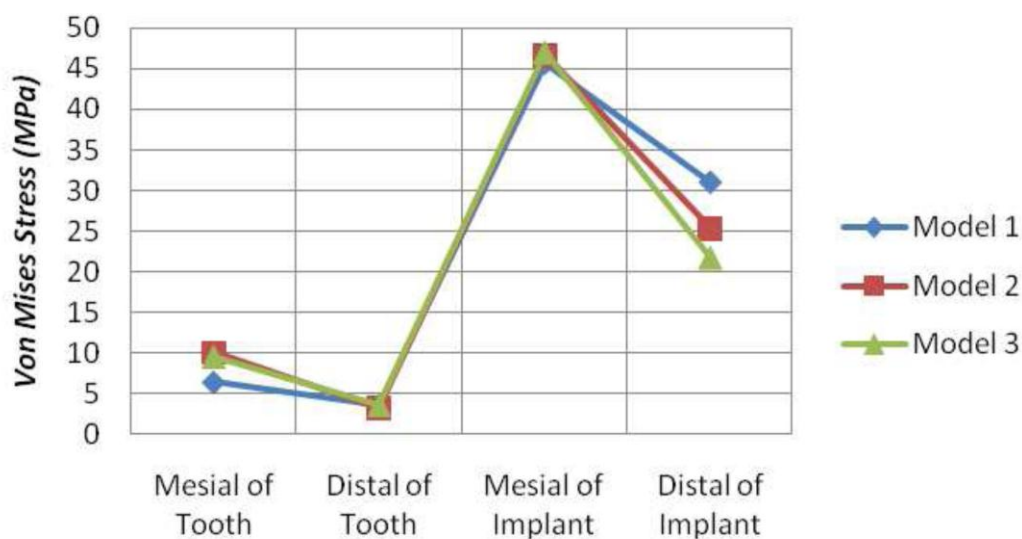


Fig 4. The equivalent Von Mises stress in the bone regions

## RESULT

### Stress at Bone Regions

The patterns of stress distribution in all three models were almost similar, but the differences were in the stress values. The maximum stress values in the mesial and distal surfaces of the crestal region of the implant bone interface was 45.62 and 31.05 MPa, respectively. The maximum stress values accumulated around the natural tooth were 6.47 MPa in the mesial and 3.68 MPa in the distal crestal surface. The equivalent Von Mises stress contours for the rigid connection configuration are shown in Figure 2A.

Model 2 (The second premolar and the implant are connected by a non-rigid attachment with the matrix connector positioned in the distal side of the second premolar):

The highest equivalent Von Mises stress values were obtained in the cortical bone region of both mesial and distal sides with values ranging between 46.69 and 25.38 MPa, respectively. The maximum stresses around the natural tooth were 10.10 MPa and 3.29 MPa in mesial and distal crestal region, respectively. The Von Mises stress contours for Model 2 are shown in Figure 2B.

Table 3. Von Mises Stresses at Critical Regions (MPa)

	Around Natural Tooth (MPa)		Around Implant Abutment (MPa)	
	Mesial	Distal	Mesial	Distal
Model 1	6.47	3.68	45.62	31.05
Model 2	10.10	3.29	46.69	25.38
Model 3	9.64	3.67	46.98	21.78

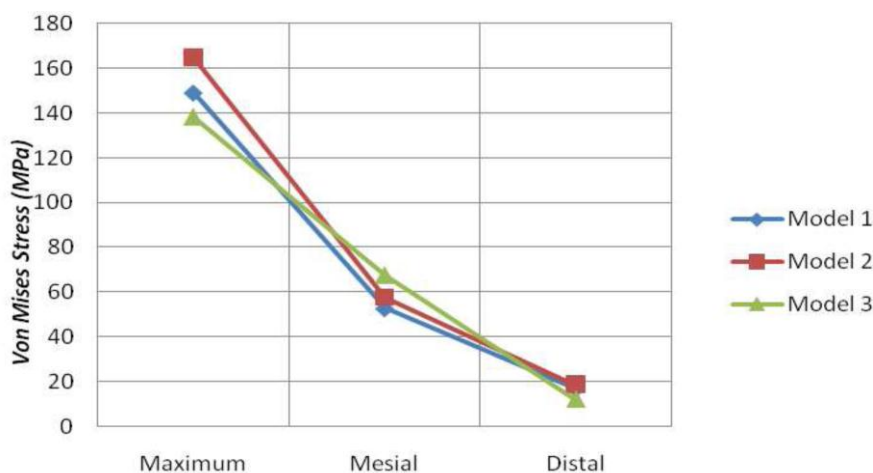


Fig 5. The Von Mises stress of the implant

Model 3 (The second premolar and the implant are connected by a non-rigid attachment with the matrix connector positioned on the mesial side of the implant): The highest equivalent Von Mises stress values were 46.98 and 21.78 MPa, respectively in the mesial and distal cortical region of the implant abutment. The stresses around the natural tooth were 9.64 and 3.67 MPa in the mesial and distal cortical region, respectively. The equivalent Von Mises stress contours for Model 3 are shown in Figure 2C. Maximum equivalent Von Mises stress values in selected critical regions of the models are summarized in Table 3.

**Stress at the Implant System**

There was no significant difference in the patterns of Von Mises stress distribution in implant system of all three models.

But, the values of stress were different in each model.

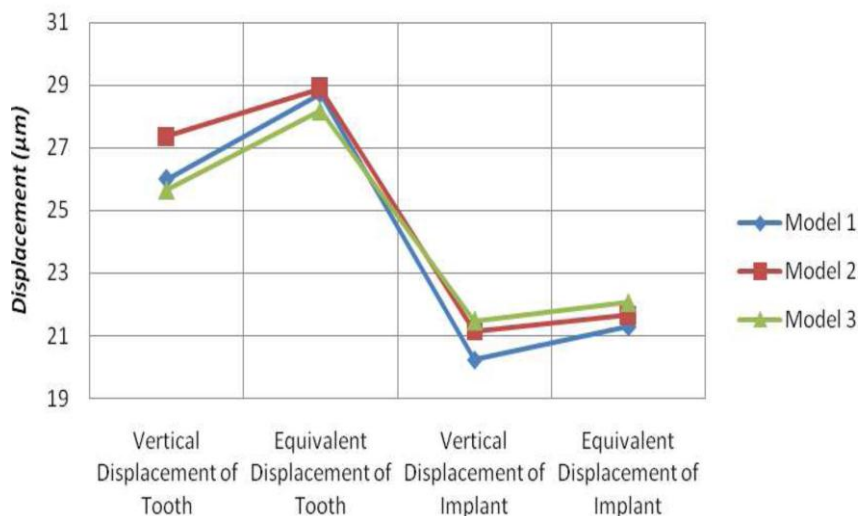
Model 1: The maximum stress value on the implant was 149.1 MPa. The maximum stresses in mesial and distal side of the implant were 52.74 and 16.88 MPa, respectively. The Von Mises stress contours of the implant for Model 1 are shown in Figure 3.

Model 2: The highest Von Mises stress was 165.0 MPa for the implant system.

Table 4. Von Mises Stresses at the Implant System (MPa)

	Maximum (MPa)	Mesial (MPa)	Distal (MPa)
Model 1	149.1	52.74	16.88
Model 2	165.0	57.66	18.70
Model 3	138.3	67.59	12.17





**Fig 6.** The displacement of the natural tooth and the implant

The maximum Von Mises stresses were 57.66 and 18.70 MPa in the mesial and distal of the implant, respectively. The Von Mises stress contours of the implant for Model 2 are shown in Figure 3.

**Model 3:** The highest equivalent Von Mises stress value was 138.3 MPa in the implant.

The maximum Von Mises stresses accumulated in the mesial and distal side of the implant were 67.59 and 12.17 MPa, respectively.

The Von Mises stress contours of the implant for Model 3 are shown in Figure 3. The maximum equivalent Von Mises stress values in all three models are summarized in Table 4.

#### **Displacement of the tooth and the Implant**

**Model 1:** The maximum vertical displacements of the natural tooth and the implant were 26.01 and 20.24 µm, respectively. In addition, the maximum equivalent displacement was 28.74 µm for the natural tooth and 21.30 µm for the implant.

**Model 2:** The highest vertical displacements obtained for the natural tooth and the implant were 27.37 and 21.15 µm, respectively.

The maximum equivalent displacements were 28.89 µm for the natural tooth and 21.66 µm for the implant.

**Model 3:** The highest vertical displacement values were 25.65 and 21.48 µm for the natural tooth and the implant, respectively. The maximum equivalent displacement was 28.19 µm for the natural tooth and 22.06 µm for the implant. The maximum vertical and equivalent displacement values for the natural tooth and the implant of the models are summarized in Table 5.

#### **DISCUSSION**

The study of biomechanics of stress loaded in dentistry has been performed widely by the 3D FEA model; however, the current investigations used in this study by the FEA program were limited by the unrealistic assumptions such as homogeneous, linear elastic and isotropic condition for the bone, tooth and periodontal ligament.

Furthermore, in this method it has been assumed that bonding of the bone and the implant are perfect. In this study, masticatory

forces were static and loaded axially relative to the occlusal plane compared with the dynamic masticatory forces, which are oblique to the occlusal surface. Consequently, the reconstruction of all the natural details can hardly be obtained.

There is some controversy about the mechanical property of the PDL. This range is about 0.01MPa to 1750 MPa [49].

The Young's modulus of the PDL in this article seems to be a generally well-accepted value and has been used in many other similar studies. So Young's modulus of the PDL was selected as 69MPa [21,24,25,34,48].

Taking these limitations into account, identical stress values with reality cannot be achieved in this study, but the differences in stress and cones of different TIFP designs could be observed. Stress increased in the mesiocervical surface of the implant in all TIFP designs. The implant movement in the alveolus is at the micron level as the rigid anchorage between the bone and the implant [26,47].

Comparing the natural tooth with the implant, intrusion may occur during mastication in natural dentition, while stress may accumulate around the implant.

Moreover, the rotational center in the implant which is at the crestal bone level is much higher than natural teeth [20,34,46]. Therefore, the cortical bone is the stress accumulation area in the implant support [34].

Formation of cortical and spongy bone structure with two different structures and different elastic moduli and rigidity make the cortical bone susceptible to stress accumulation in this area [20,34]. A similar outcome was achieved by Mish and Ismail regarding 3D FEA results [32]. Moreover, Melo et al. evaluated the effect of NRCs in the amount of stress accumulated in the surrounding bone of TIFPs and they did not observe any reduction of stress [33]. In Menicucci et al. in vitro studies, the deteriorating effect was found in the static load compared with the transitional one.

In addition, they stated that periodontal ligament is the pivotal factor in the distribution of applied force between the tooth and the implant with rigid connection [23].

In terms of location of the NRC in the TIFP designs, Bechelli advocates the implant support sides. Physiological movement of the natural tooth, protection from torque effect and equal distribution of force on the implant and the tooth are the advantages of Bechelli's supported design [13].

Burak et al. investigated these three models by 2D FEA and photoelastic analysis and stated that the stress on implant support was decreased more in model 3 than models 1 and 2. These last two models had the same stress distribution pattern [34]. This study is contrary to Burak's finding and the stress distribution patterns were similar in the three models.

**Table5.** Vertical and equivalent displacement at the natural tooth and the implant ( $\mu\text{m}$ )

	Natural Tooth ( $\mu\text{m}$ )		Implant ( $\mu\text{m}$ )	
	Vertical displacement	Equivalent displacement	Vertical displacement	Equivalent displacement
Model 1	26.01	28.74	20.24	21.30
Model 2	27.37	28.89	21.15	21.66
Model 3	25.65	28.19	21.48	22.06

Stress distribution in the mesiocervical area was much greater than the distocervical in the three models and it was greater in models 2 and 3 than model 1 (Fig. 4).

It seems that the occlusal load of TIFP did not distribute equally in the supportive area. Stress distribution in model 3 was less than model 1 and 2 as the attachment was on the mesial side of the implant.

Implant stress distribution in model 2 was greater than the two other systems, as the pontic was cantilevered (Fig. 5).

Implant mobility was greater in model 3 and lesser in model 1, so placement of the attachment implant may have an important role (Fig. 6).

## CONCLUSION

Within the limitations of this study, the following conclusions are drawn:

- 1- According to this 3D FEA, there is no difference in stress distribution of the implant bone support in the 3 models.
- 2- Displacement of the implant in model 3 was greater than the two other models.
- 3- There was less stress distribution in implant system in model 3 than the two other models.
- 4- Different TIFP designs did not affect the stress accumulation around the natural teeth.

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